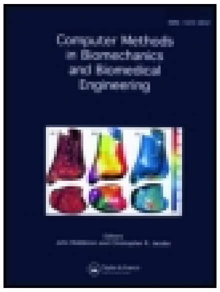

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The application of musculoskeletal modeling to investigate gender bias in non-contact ACL injury rate during single-leg landings

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The central tenet of this study was to develop, validate and apply various individualised 3D musculoskeletal models of the human body for application to single-leg landings over increasing vertical heights and horizontal distances. While contributing to an understanding of whether gender differences explain the higher rate of non-contact anterior cruciate ligament (ACL) injuries among females, this study also correlated various musculoskeletal variables significantly impacted by gender, height and/or distance and their interactions with two ACL injury-risk predictor variables; peak vertical ground reaction force (VGRF) and peak proximal tibia anterior shear force (PTASF). Kinematic, kinetic and electromyography data of three male and three female subjects were measured. Results revealed no significant gender differences in the musculoskeletal variables tested except peak VGRF ($p = 0.039$) and hip axial compressive force ($p = 0.032$). The quadriceps and the gastrocnemius muscle forces had significant correlations with peak PTASF ($r = 0.85$, $p < 0.05$ and $r = -0.88$, $p < 0.05$, respectively). Furthermore, hamstring muscle force was significantly correlated with peak VGRF ($r = -0.90$, $p < 0.05$). The ankle flexion angle was significantly correlated with peak PTASF ($r = -0.82$, $p < 0.05$). Our findings indicate that compared to males, females did not exhibit significantly different muscle forces, or ankle, knee and hip flexion angles during single-leg landings that would explain the gender bias in non-contact ACL injury rate. Our results also suggest that higher quadriceps muscle force increases the risk, while higher hamstring and gastrocnemius muscle forces as well as ankle flexion angle reduce the risk of non-contact ACL injury.

Keywords: non-contact ACL injury; muscle forces; muscle activity; joint reaction forces; proximal tibia anterior shear force

Abbreviations: ACL, anterior cruciate ligament; ATT, anterior tibial translation; MSM, musculoskeletal model; GRFs, ground reaction forces; VGRF, vertical ground reaction force; PGRF, posterior ground reaction force; PTASF, proximal tibia anterior shear force; EMG, electromyography; AMS, AnyBody Modeling System

1. Introduction

Single-leg landing is a common task performed from varying vertical heights and horizontal distances during sports. The literature indicates that the highest incidence of non-contact anterior cruciate ligament (ACL) injury occurs during single-leg landing sports such as basketball, soccer and team handball (Kirkendall and Garrett 2000; Paul et al. 2003; Renstrom et al. 2008; Boden et al. 2009). In spite of this, little is known about muscle response and loading when ACL injury occurs during single-leg landings. This may be attributed to the difficulties associated with measuring joint reaction and muscle forces *in vivo*, as well as the level of effort required to develop and simulate musculoskeletal models (MSMs).

There are many single-leg landing studies in the literature (Self and Paine 2001; Lephart et al. 2002; Fagenbaum and Darling 2003; Hargrave et al. 2003; Ford et al. 2006; Russell et al. 2006; Nagano et al. 2007; Pappas et al. 2007; Schmitz et al. 2007; Lawrence et al. 2008; Kiriyama et al. 2009; Shimokochi et al. 2009; Yeow et al. 2010; Laughlin et al. 2011), but none had utilised a MSM to investigate single-leg landings over increasing vertical heights and horizontal distances. More importantly, none of these studies examined the relationships among height and distance of landing, ground reaction forces (GRFs), joint reaction forces, muscle forces, joint kinematics and risk of non-contact ACL injury. McLean et al. (2003, 2004, 2005), Shelburne and Pandy (2002) and Lloyd and Besier (2003) utilised a MSM to determine the forces in the muscles during activities implicated to cause non-contact ACL injury. For many of these studies, the aim was to determine the joint reaction or muscles forces during activities implicated to cause ACL injuries. The McLean and Lloyd studies (Lloyd and Besier 2003; McLean et al. 2003, 2004, 2005) focused on side-step cutting as a non-contact ACL injury mechanism, while the work of Pandy's research group (Pandy and Shelburne 1997; Shelburne and Pandy 1997, 2002; Anderson and Pandy 1999; Pflum et al. 2004) focused on double-leg landings. Even though a recent study (Laughlin et al. 2011) investigated single-leg landings

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using musculoskeletal modeling, they did not address the effect of height, distance or gender on the risk of ACL injury. To the authors' best knowledge, this study for the first time investigated the main effect of increasing horizontal distance as well as the interaction of vertical height, horizontal distance and gender on single-leg landing biomechanics, and further relate these findings to risk of non-contact ACL injury. Also for the first time, this study reported data on lower extremity joint reaction and muscle forces during single-leg landing over increasing vertical heights and horizontal distances.

The ability of body kinematics or lower extremity muscles to attenuate the GRFs upon landing on a single-leg may enable us to better prevent ACL injuries through improved biomechanical function. If these impact forces cannot be dissipated by joint motion or joint and muscle forces, the ACL may be forced to carry the brunt of the load and increase the risk of injury. Two studies (Boden et al. 2009; Podraza and White 2010) showed that a lack of absorption of GRFs at landing may be a factor in ACL injury. In addition, an *in vivo* study (Cerulli et al. 2003) demonstrated that for a male subject hopping and landing on one leg, peak ACL strain occurred at peak VGRF, suggesting that peak VGRF may be a likely predictor of risk of non-contact ACL injuries. Other studies (Malinzak et al. 2001; Chappell et al. 2002; Madigan and Pidcoe 2003) confirmed that peak VGRF may amplify internal joint loads that may cause ACL injury if not sufficiently attenuated by the musculoskeletal system. Therefore, this study uses peak VGRF as an ACL injury risk predictor variable. Peak PTASF was also selected as an ACL injury risk predictor variable in this study, given that an increase in this force can lead to an increase in anterior tibial translation (ATT), which increases ACL loading (Fleming et al. 2001a, 2001b; DeMorat et al. 2004; Withrow et al. 2006; Yu et al. 2006; Yu and Garrett 2007). Given this, our study investigated the relationship between these two non-contact ACL injury risk predictor variables and lower extremity joint reaction forces, muscle forces as well as lower extremity kinematics.

To the authors' best knowledge, there are no *in vivo* or musculoskeletal modeling studies reporting joint reaction and muscle forces between genders during single-leg landings over increasing vertical heights and horizontal distances. The objective of this study is threefold: firstly, develop and validate 3D individualised MSMs of the human body for application to single-leg landings; secondly, determine gender differences with respect to joint reaction and muscle forces as well as lower extremity kinematics during single-leg landings and finally, correlate the dependent variables significantly impacted by the main effect and interaction of gender, vertical height, and horizontal distance with two possible non-contact ACL injury risk predictor variables. We hypothesise that females will land with significantly greater peak VGRFs compared to males. We also hypothesise that males and females would demonstrate significantly different lower extremity joint reaction and muscle forces which could explain the higher number of ACL injuries in females.

2. Method

2.1 Experimental procedure

Three male recreational athletes with mean (SD) age of 22.8 (1.60) years, heights of 1.80 (0.03) m and masses of 67.28 (3.52) kg, and three female recreational athletes age of 21.6 (1.20) years, heights of 1.71 (0.02) m and masses of 64.71 (2.33) kg were recruited from the university population. Males and females were weight and height matched as closely as possible. None of the participants reported any musculoskeletal or ligamentous injuries to the lower extremity at the time of participation. Prior to data collection, each participant gave informed consent as stipulated by the university ethics review board. Subjects' ages, masses and heights were recorded. The dominant leg was established as the leg used by the subject to kick a ball. All participants wore identical shoes (running shoe, model BY004, ASICS America Corporation, Irvine, CA, USA) throughout data collection so as to mitigate variability. Retroreflective markers were affixed with double-sided tape using a customised marker protocol as shown in Figure 1. A motion capture system (Vicon MX, Oxford Metrics, UK) consisting of seven infrared video cameras collected marker trajectories at a sampling rate of 250 Hz. A force plate (Kistler type 9281B, Winterthur, Switzerland) measured GRFs data at sampling rate of 1000 Hz. A Bortec AMT-8 EMG System (Bortec Biomedical Ltd, Calgary, Canada) measured surface electrode electromyography (EMG) of eight major muscles of the dominant leg (biceps femoris, vastus medialis, gastrocnemius, gluteus maximus, medial semitendinosus, tibialis anterior, soleus and rectus femoris) at a sampling rate of 1000 Hz. Motion, force plate and EMG data were time synchronised. Subjects began the landing task by standing on an elevated deck with hands placed on their iliac crests, legs shoulder width apart and the toes of both legs aligned with the edge of the deck. Subjects were then instructed to stand on their dominant leg alone, jump forward and land as naturally as possible with the dominant foot centred on the force platform. The subjects were asked to keep their hands on their iliac crests throughout the trials to reduce any variability from swinging arms. The participants were instructed to perform the task from the elevated deck of increasing vertical heights (20, 40 and 60 cm) that was placed at increasing horizontal distances (30, 50 and 70 cm) from the edge of the force platform. The nine different landing configurations (combination of landing height and distance) tested were h20d30, h20d50, h20d70, h40d30, h40d50, h40d70, h60d30, h60d50 and h60d70, where *h* represents the vertical landing height and *d* represents the

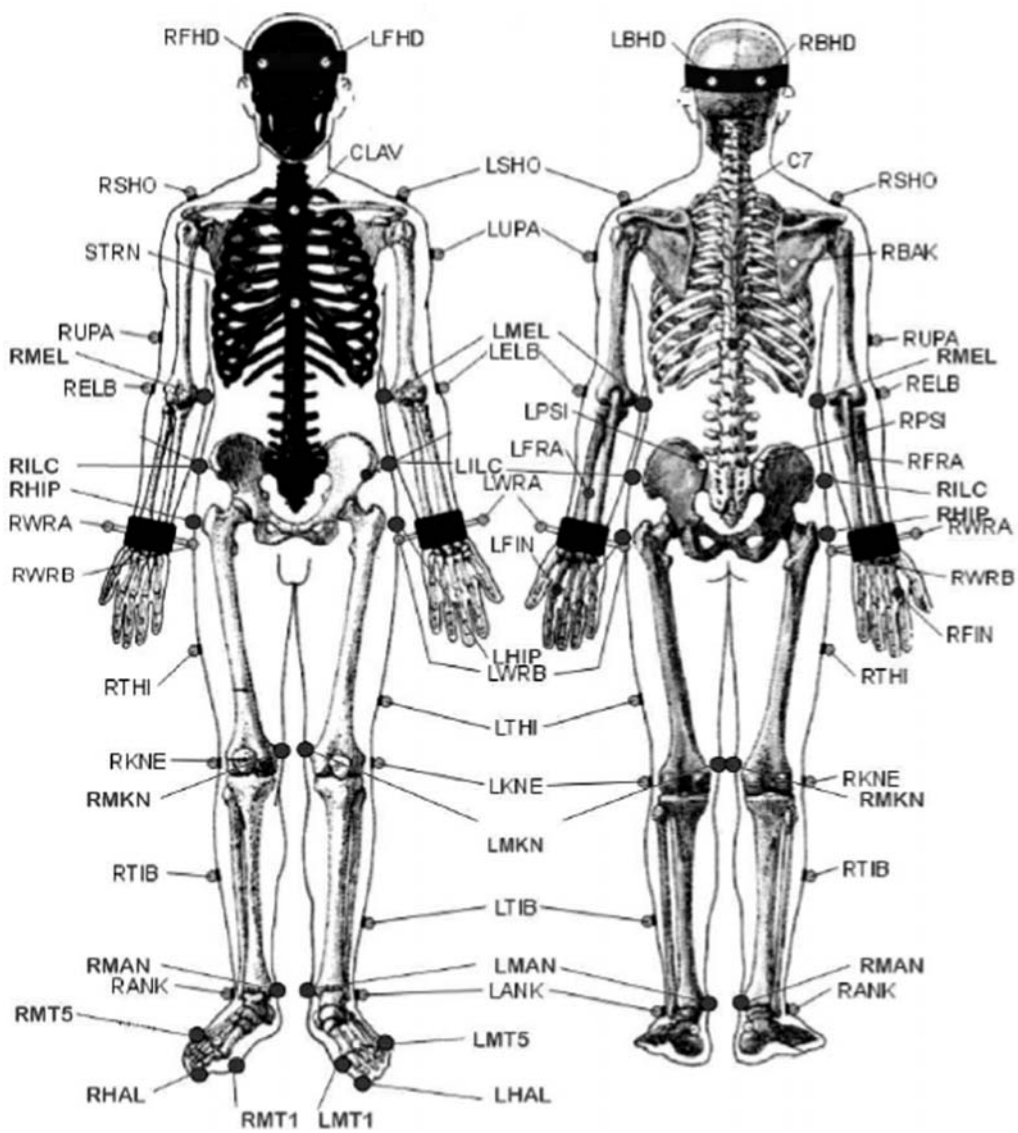

Figure 1. Customised marker set used in this study. Adapted from Oxford Metrics Plug-in-gait marker placement document.

horizontal landing distance. The numbers after *h* and *d* refer to the heights and distances, respectively, in centimetres. The sequence of landing configuration was randomised to reduce learning effects.

2.2 Model development and validation

The AnyBody Modeling System (AMS) software (AnyBody Technology A/S, Aalborg, Denmark) is based on inverse dynamics analysis (IDA) as well as optimisation principles, and was used to develop, validate and simulate the motion of the individualised MSMs used for this study. Inverse dynamics analysis was used to determine the unknown joint reaction and muscle forces from the known motion. Given there are more muscles than there are degrees of freedom, the redundancy posed by this muscle recruitment problem indicates that there is no unique solution, and as such, the problem is formulated as an optimisation problem. In this optimisation problem, the objective function is geared towards minimising the maximum muscle activity subject to equilibrium constraints and positive muscle force constraints (i.e. muscles can only pull). This approach arguably provides more detailed information than rigid body models solved using classical IDA that cannot calculate muscle forces and consequently cannot be used to assess joint reaction forces. Muscle inclusion is important, given that during motion the externally applied forces create moments about the joints that have to be balanced by muscles. Since the moment arms of muscles are much smaller than moment arms of

externally applied forces, the muscles have to pull relatively more on the bones to obtain equilibrium. So the muscles' contributions to the joint reaction forces are significant, and therefore rigid body models using classical IDA devoid of muscles grossly underestimate joint reaction forces. Furthermore, rigid body models using classical IDA often ignore gravity and inertia; that is, they use quasi-static assumption. Details of the mathematical and mechanical methods of the AMS software are described in the literature (Rasmussen et al. 2001; Damsgaard et al. 2006).

The development of a MSM from the ground up is a very time consuming and complex endeavour. To mitigate these challenges, a publicly shared model repository (AnyBody Managed Model Repository V1.2) was created by the AnyBody research group. The GaitFullBody MSM was extracted from this repository, modified for this study, individualised for each subject, validated and then applied to the single-leg landing trials. The GaitFullBody MSM is based on an anthropometric data set by Klein Horsman et al. (2007). The motion and force plate data at each landing configuration measured experimentally were used as inputs to the AMS software. The GaitFullBody model was combined with each subject's single-leg landing trials to individualise the model. More specifically, the model was scaled to fit the dimensions of the subject, given the retroreflective marker positions relative to each other via an optimisation routine. This optimisation method simultaneously optimises the model scaling, local marker coordinates and model motion during a dynamic trial so as to minimise the differences between model markers and the measured marker trajectories (Andersen et al. 2010). Once optimised kinematics were derived (determined by mean absolute error <0.5 between model kinematics and experimental kinematics), IDA was then performed with a minimum/maximum recruitment solver to address the redundancy challenge (Rasmussen et al. 2001).

The GaitFullBody model included both the major upper and lower extremity joints but only the muscles of the lower extremity in order to reduce computational time. The hips were modelled as a spherical joint, the knee as a revolute joint and ankle as a universal joint. The GaitFullBody model was actuated by a total of 70 muscle-tendon units-35 muscles on each leg. The model utilised a simple muscle model that did not consider the force-length and force-velocity relationships as well as the passive stiffness of the muscles as the Hill's muscle model does. The muscle attachment sites and geometries were scaled in accordance with a linear geometry scaling law (Equation (1); Rasmussen 2005):

$$s = Sp + t, \tag{1}$$

where s is the scaled point, S is the scaling matrix, p is the original point and t is the translation. A length-mass-fat scaling law was used to scale soft tissues, taking into account each subject's body weight, body height and segment lengths. For greater details on the scaling algorithm used for this study, interested readers can consult the work of Rasmussen (2005).

Validation of MSMs is a very challenging task, since it is difficult and sometimes impossible to measure muscle or joint reaction forces *in vivo*. As a paradox, MSMs are developed to determine these forces. To corroborate the individualised MSMs, two different approaches were taken; firstly, we compared the predicted and the measured muscle activations of eight muscles. The predicted muscle activity should occur at approximately the same time as the measured muscle activity. This information is presented in an on–off timing curve (Figure 2) that shows the time when the measured (fat line) and predicted (thin lines) muscles goes above (turns on) and below (turns off) a threshold of muscle activity during a single-leg landing task. A threshold of 20% was used. A threshold of 20% was chosen to provide a reasonable number of data points to compare the predicted and the measured muscle activities during landing. As gleaned from Figure 2, the measured muscle activity tends to occur prior to the predicted muscle activity. This is as expected, given that the latency of a muscle's response to stimulus is not captured in the individualised MSMs. Furthermore, at each landing configuration, the predicted and measured muscle activities can be represented by a linear envelop. The average Pearson's product–moment correlation (PPMC) between measured and predicted muscle activities for eight muscles and across all landing configurations was 0.58 for a subject. The average PPMC for all six subjects between measured and predicted muscle activities was 0.52. The trend of the measured and predicted muscle activation curves (see sample in Figure 3) looks similar but with sudden spikes in predicted data. The reason for the sudden changes in the predicted muscle activation is that the inverse dynamics calculation at every time step in the AMS software is completely independent of the other.

Secondly, the knee joint forces and moments measured *in vivo* from telemetered joint replacements during walking gait and reported in the literature for both males and females (Lu et al. 1997, 1998; Taylor et al. 1998; Taylor and Walker 2001; D'Lima et al. 2005; Heinlein et al. 2009; Kutzner et al. 2010) were compared with predicted knee joint forces and moments. The authors recognise that the best way to validate the outputs from the MSMs during single-leg landings would be to compare them to *in vivo* data obtained for the identical task. To the authors' best knowledge, no *in vivo* data on the lower extremity joint forces or moments during single-leg landing exist to date. To aid model validation and increase confidence in our MSM, the model was driven with walking gait data collected from the same six subjects that performed the single-leg landing tasks. The walking gait trials were

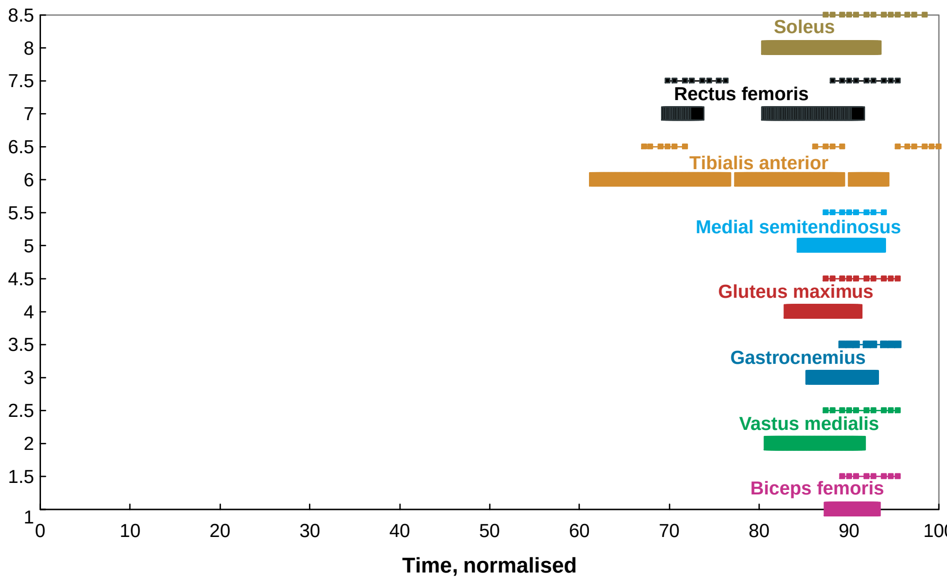

Figure 2. Measured EMG timing (fat line) compared with predicted muscle activity timing (thin line) during single-leg landing.

conducted before the single-leg landing trials. The subjects were asked to walk along a 15-m raised platform at a speed that closely matched the average speed of participants used in the literature while motion and force plate data were collected. A comparison of the predicted and *in vivo* knee joint forces and moments reported in the literature is provided in Table 1. Given our MSM predictions reported in Table 1 and recognising variability in body anthropometry between studies (some subjects from *in vivo* studies were at least 300 N heavier and 10 cm taller than the subjects from our population), it appears that the applied individualised MSMs tend to provide reasonable estimates of the internal knee joint moments and medial–lateral joint reaction forces while systematically overestimating the anterior–posterior and axial joint reaction forces. The latter can be controlled in a MSM by adjustment of muscle moment arms, but in the interest of reproducibility, it was elected to not pursue this option.

2.3 *Model application, data reduction and analysis*

Upon completion of the validation process outlined above, simulation of the single-leg landing tasks for all subjects was conducted for the nine landing configurations using the validated MSM. An example of the individualised validated MSM is provided in Figure 4. Figure 4 presents a validated individualised MSM being applied at a single-leg landing configuration. The time at which peak VGRF occurred was used to determine the joint reaction and muscle forces, as well as joint kinematics. All kinematic and force plate data were low-pass filtered using a second-order bidirectional Butterworth filter at 6 and 25 Hz, respectively. These data were resampled using linear interpolation, synchronised using events and then ensemble averaged using Matlab (The MathWorks, Inc., Natick, MA, USA). The EMG signals were first rectified and then low-pass filtered at a cut-off frequency of 6 Hz using a critically-damped, zero-lag filter. The processed

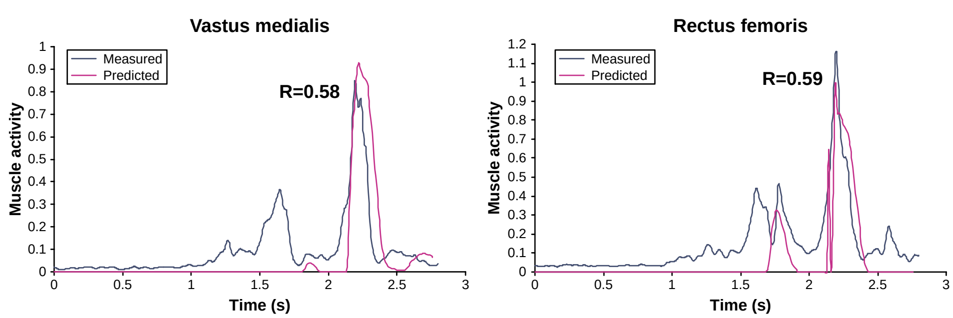

Figure 3. Sample plots showing trends in measured and predicted muscle activities for the vastus medialis and rectus femoris.

Table 1. Comparison of predicted and published *in vivo* knee joint reaction forces during walking gait.

	Predicted knee joint forces and moments by AnyBody MSM during walking gait	*In vivo* knee joint forces and moments from the literature during walking gait
F_z (%BW) (axial direction)	376.50	223–300 (Heinlein et al. 2009; Kutzner et al. 2010); 250 (Taylor et al. 1998); 320 (Lu et al. 1997); 280 (Lu et al. 1998; Taylor and Walker 2001; D'Lima et al. 2005)
F_y (%BW) (medial–lateral direction)	30.78	17–46 (Heinlein et al. 2009; Kutzner et al. 2010)
F_x (%BW) (anterior–posterior direction)	40.50	10–36 (Heinlein et al. 2009; Kutzner et al. 2010)
M_z (%BW × BH)	0.91	0.37–1.23 (Heinlein et al. 2009; Kutzner et al. 2010)
M_y (%BW × BH)	1.76	0.52–2.57 (Taylor and Walker 2001)

Note: Forces are normalised to bodyweight (BW) and moments to bodyweight × body height (BW × BH).

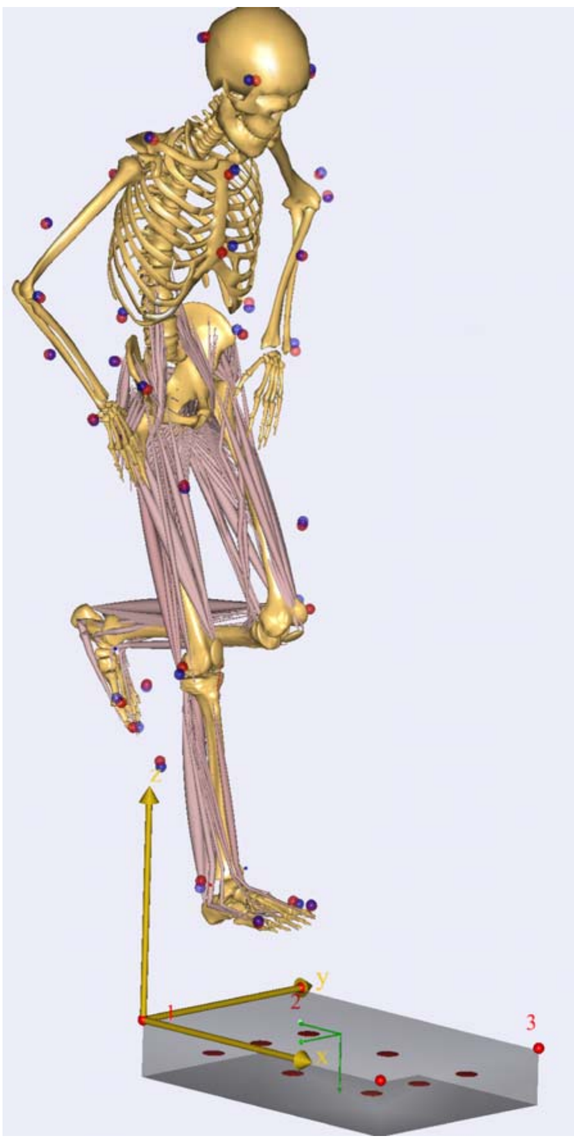

Figure 4. Validated 3D full body MSM of the human body applied to a single-leg landing task.

EMG signals were normalised to the maximum measured EMG value for that muscle over all the measurements.

2.4 Statistical analysis

Multiple three-way mixed design with repeated-measures ANOVAs (two within subject and one between subject) were conducted to test the main effects and interactions of three independent variables (height, distance and gender) with various musculoskeletal-dependent variables. In this study, vertical height and horizontal distance were the within-subject factors, gender was the between-subjects factor, and the dependent variables were joint reaction and muscle forces, as well as ankle, knee and hip flexion angle. Follow-up testing entailed PPMCs measured to determine the relationship between the two possible ACL injury risk predictor variables, namely peak VGRF and peak PTASF, and any of the dependent variables from ANOVA analyses significantly affected by height, distance or gender. Statistical analyses were conducted in SPSS (SPSS for Windows, Release 11.5.0), in which a significance level of $\alpha = 0.05$ was utilised.

3. Results

This section presents the outputs from the musculoskeletal modeling simulations that tested the main effects and interactions of height, distance and gender on single-leg landing. The selected outputs from the MSMs, namely, joint reaction forces, muscle forces, as well as ankle, knee and hip flexion angle are presented for the nine different landing configurations. Figure 5 shows the ensemble averages of VGRF, PGRF and PTASF histories (Figure 5(a)–(c) for males and Figure 5(d)–(f) for females, respectively), as well as quadriceps, hamstrings and gastrocnemius muscle force histories (Figure 5(g)–(i) for males and Figure 5(j)–(l) for females, respectively) at the nine landing configurations.

The key findings from the separate ANOVAs conducted are provided in Table 2. The results showed a significant main effect for gender $F(1,4) = 11.64$,

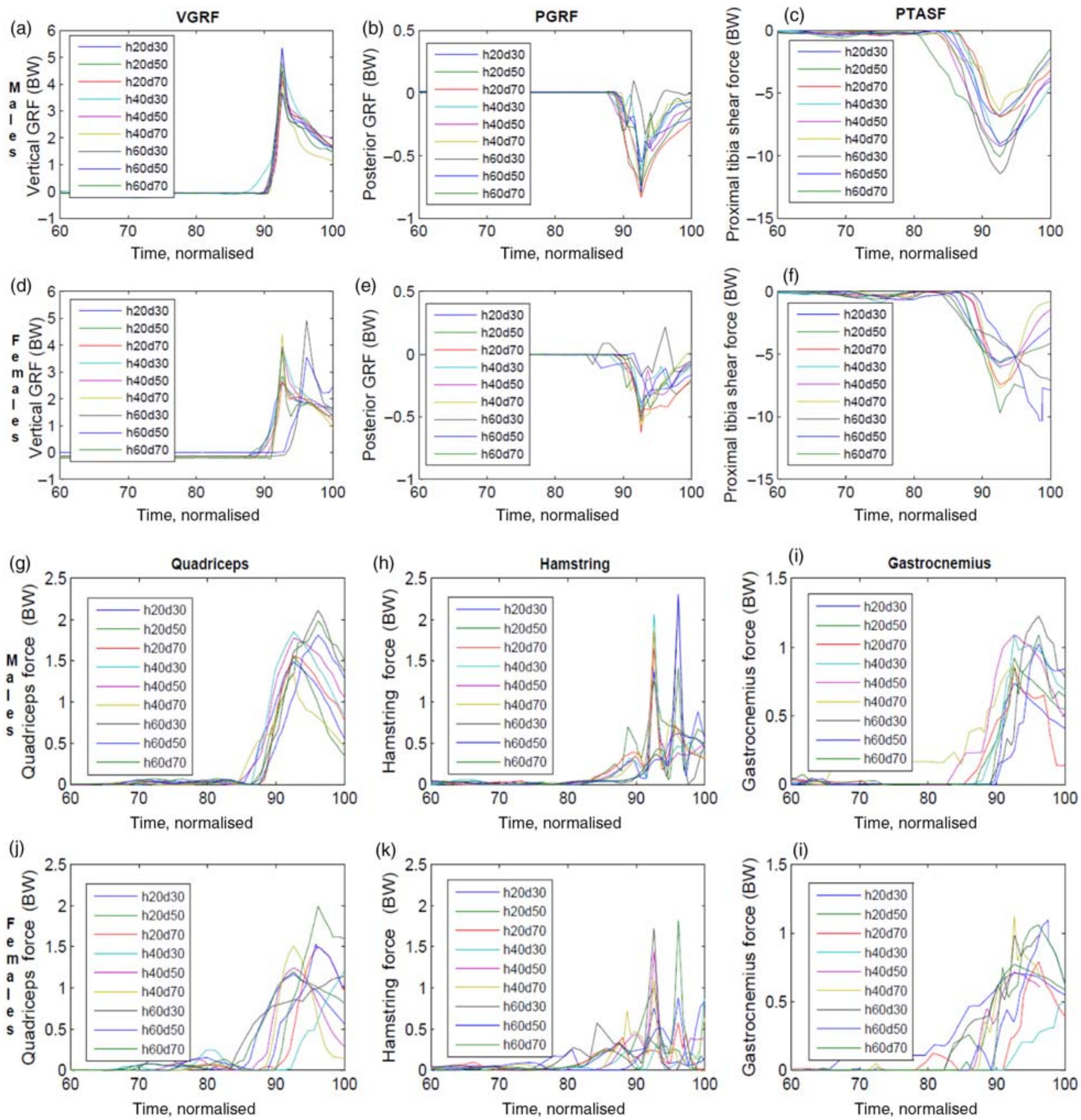

Figure 5. The ensemble averages of VGRF, PGRF and PTASF histories as well as quadriceps, hamstring and gastrocnemius muscle force histories during the nine single-leg landing configurations for males (first and third rows, respectively) and females (second and fourth rows, respectively).

$p < 0.05$, partial $\eta^2 = 0.744$ with peak VGRF and $F(1,4) = 10.515$, $p < 0.05$, partial $\eta^2 = 0.724$ with hip axial compressive force. Here partial η^2 is a measure of effect size with a larger value indicating the more variance the effect explains in the dependent variable. Of the dependent variables studied, females had significantly lower peak VGRF and hip axial compressive forces compared to males (Table 3(a) and (b)).

There was a significant distance × gender interaction $F(2,8) = 6.85$, $p < 0.05$ and partial $\eta^2 = 0.63$ for quadriceps muscle force. From descriptive statistics (Table 3(a) and (b)), the largest difference in quadriceps muscle force between genders occurred at $d = 30$ cm and $d = 50$ cm; it is these differences that are significant. It was interesting to find a significant height × distance interaction on both ACL injury-risk predictor variables, suggesting perhaps that horizontal distance of landing, often not considered in

Table 2. ANOVA summary showing interactions and the main effects observed.

	Dependent variables										
	Peak VGRF	Peak PTASF	Ankle axial compressive force	Knee axial compressive force	Hip axial compressive force	Quadriceps muscle force	Hamstring muscle force	Gastrocnemius muscle force	Ankle flexion angle	Knee flexion angle	Hip flexion angle
Height	**	***		**	***	*	*	*		**	*
Distance		*							*	**	
Height × gender											
Distance × gender						*					
Height × distance	*	*									
Height × distance × gender											
Gender	*				*						

*$p < 0.05$; **$p < 0.01$; ***$p < 0.005$.

many studies, may be an important variable in single-leg landing biomechanics research. Results revealed no significant height × distance × gender or height × gender interactions (Table 2). ANOVA analysis also showed significant main effects of both height and distance with many musculoskeletal variables (Table 2), and descriptive statistics of these variables are shown in Table 3(a) and (b). Given that among the dependent variables tested, only two variables were significantly different between genders, the data for the male and female subjects were pooled together for the subsequent statistical analysis.

Follow-up statistical testing entailed PPMCs measured to determine the relationship among ACL injury-risk predictor variables and the variables from ANOVA analysis significantly affected by the main effect of height or distance (Table 4). From Table 4, one can glean some significant associations between the two ACL injury-risk predictor variables and the variables affected by the main effects of landing height and distance. Finally, Figure 6(a)–(c) shows the normalised muscle forces of males (blue) and females (yellow) for the quadriceps, hamstrings and gastrocnemius at the nine landing configurations.

4. Discussion and conclusion

To assess the risk of non-contact ACL injuries during single-leg landings, as well as to suggest prevention strategies to reduce the likelihood of injuries during such a task, it is imperative to understand the effects of increasing vertical height and horizontal distance of landing on underlying musculoskeletal variables. The literature reviewed clearly identifies the need for MSMs applied to single-leg landings.

This study showed that single-leg landings did not produce the characteristic bimodal VGRF curve commonly reported for double-leg landings (Dufek and Bates 1990; McNitt-Gray 1993). It appears that the demands of single-leg landings from this study resulted in a rapid increase in GRFs with a single peak (Figure 5(a) and 5 (d)), which is in agreement with the literature (Hargrave et al. 2003). This finding is important as it elucidates the unique nature of single-leg landing studies whose findings cannot be compared with double-leg landing studies.

Given this study is the first to report joint reaction and muscle forces during single-leg landings over increasing vertical heights and horizontal distances of landing, there are no studies to draw direct comparisons, so we cannot confirm that the joint reaction and muscle forces predicted by the MSMs for the various single-leg landing configurations are accurate. We found no gender differences in ankle, knee or hip flexion angle during single-leg landing. This is in agreement with earlier research that showed no difference in ankle flexion angle (Schmitz et al. 2007), knee flexion angle (Nagano et al. 2007; Kiriyama et al. 2009) or hip flexion angle (Lephart et al. 2002; Pappas

Table 3. Descriptive statistics (mean ± SD) of risk predictor and dependent variables for males (a) and females (b).

Dependent variables	h20d30	h20d50	h20d70	h40d30	h40d50	h40d70	h60d30	h60d50	h60d70
(a)									
				Males (n = 3)					
Peak VGRF (BW)	3.59 ± 0.24	3.69 ± 0.19	4.11 ± 0.47	4.80 ± 0.2	4.36 ± 0.21	4.8 ± 0.72	5.37 ± 0.62	5.37 ± 0.94	4.52 ± 0.55
Peak PTASF (BW)	−6.84 ± 2.65	−6.99 ± 2.23	−6.92 ± 2.24	−9.34 ± 3.06	−9.29 ± 1.46	−6.47 ± 2.47	−11.52 ± 3.01	−9.11 ± 2.53	−10.15 ± 1.86
Ankle axial compressive force (BW)	11.53 ± 3.18	12.85 ± 1.98	12.58 ± 1.26	14.69 ± 3.07	14.02 ± 2.47	11.55 ± 1.66	14.77 ± 3.39	13.31 ± 3.52	13.38 ± 3.08
Knee axial compressive force (BW)	9.44 ± 0.49	10.49 ± 0.78	10.78 ± 1.44	13.31 ± 3.54	11.35 ± 0.41	11.45 ± 1.38	14.56 ± 2.63	13.72 ± 2.38	11.81 ± 1.45
Hip axial compressive force (BW)	11.36 ± 1.05	11.54 ± 1.01	11.01 ± 0.39	16.4 ± 1.83	16.17 ± 3.06	14.93 ± 3.87	18.7 ± 4.21	18.61 ± 1.58	13.24 ± 1.19
Quadriceps muscle force (BW)	1.50 ± 0.50	1.57 ± 0.50	1.58 ± 0.44	1.58 ± 0.44	1.79 ± 0.39	1.34 ± 0.53	2.13 ± 0.58	1.83 ± 0.49	2.00 ± 0.50
Hamstrings muscle force (BW)	1.37 ± 0.16	1.25 ± 0.28	1.62 ± 0.90	2.07 ± 0.37	1.84 ± 0.76	1.88 ± 0.51	2.30 ± 0.50	2.31 ± 0.21	1.40 ± 0.43
Gastrocnemius muscle force (BW)	0.74 ± 0.37	0.93 ± 0.22	0.86 ± 0.15	1.10 ± 0.33	1.09 ± 0.23	0.79 ± 0.15	1.23 ± 0.22	1.03 ± 0.26	1.10 ± 0.28
Ankle flexion angle (deg)	48.09 ± 2.58	52.56 ± 5.18	56.55 ± 7.79	51.47 ± 2.47	55.14 ± 5.18	55.75 ± 5.77	51.97 ± 6.49	53.97 ± 4.87	57.71 ± 6.85
Knee flexion angle (deg)	54.67 ± 4.61	54.77 ± 2.27	59.47 ± 6.48	60.54 ± 0.88	67.41 ± 10.19	72.57 ± 8.62	71.23 ± 8.77	74.67 ± 11.26	80.21 ± 10.46
Hip flexion angle (deg)	19.27 ± 5.97	22.52 ± 3.92	30.92 ± 1.81	26.51 ± 2.90	27.10 ± 3.06	27.19 ± 2.77	31.07 ± 8.03	35.96 ± 3.03	38.90 ± 5.66
(b)									
				Females (n = 3)					
Peak VGRF (BW)	2.69 ± 0.44	2.97 ± 0.35	2.75 ± 0.42	4.09 ± 0.39	4.07 ± 0.29	4.59 ± 0.49	5.14 ± 1.54	4.04 ± 0.46	4.25 ± 0.66
Peak PTASF (BW)	−5.90 ± 1.70	−5.82 ± 1.86	−7.71 ± 1.77	−7.27 ± 2.14	−6.48 ± 1.83	−7.96 ± 2.16	−9.77 ± 2.55	−8.08 ± 2.78	−6.45 ± 2.55
Ankle axial compressive force (BW)	9.87 ± 3.34	9.67 ± 2.74	10.82 ± 3.65	13.36 ± 1.50	11.58 ± 1.46	13.00 ± 1.81	12.44 ± 3.57	13.20 ± 5.14	12.76 ± 5.61
Knee axial compressive force (BW)	7.71 ± 2.27	7.16 ± 2.27	7.79 ± 2.85	12.09 ± 2.49	10.66 ± 1.28	10.28 ± 0.59	12.51 ± 3.22	10.20 ± 2.21	10.5 ± 2.29
Hip axial compressive force (BW)	8.18 ± 1.57	9.62 ± 1.17	8.35 ± 0.62	13.80 ± 3.91	13.51 ± 3.75	11.64 ± 0.82	13.92 ± 4.55	9.68 ± 0.71	14.62 ± 3.2
Quadriceps muscle force (BW)	1.21 ± 0.31	1.23 ± 0.44	1.55 ± 0.36	1.24 ± 0.56	1.28 ± 0.37	1.56 ± 0.41	1.22 ± 0.86	1.53 ± 0.34	1.93 ± 0.74
Hamstrings muscle force (BW)	1.03 ± 0.30	1.08 ± 0.27	0.92 ± 0.08	1.61 ± 1.04	1.78 ± 0.83	1.49 ± 0.15	2.18 ± 0.97	1.22 ± 0.05	1.92 ± 0.49
Gastrocnemius muscle force (BW)	0.77 ± 0.31	0.80 ± 0.28	0.82 ± 0.28	0.69 ± 0.24	0.74 ± 0.28	1.16 ± 0.06	1.19 ± 0.36	0.92 ± 0.14	1.10 ± 0.42
Ankle flexion angle (deg)	49.92 ± 5.92	48.05 ± 6.37	53.30 ± 6.68	50.19 ± 12.01	51.54 ± 8.61	53.73 ± 6.60	50.47 ± 8.37	50.75 ± 8.40	51.87 ± 1.77
Knee flexion angle (deg)	47.21 ± 7.35	56.38 ± 11.22	61.04 ± 4.47	52.66 ± 11.24	59.66 ± 11.16	49.67 ± 12.06	65.05 ± 2.65	62.58 ± 6.75	73.96 ± 5.67
Hip flexion angle (deg)	24.38 ± 4.14	27.68 ± 7.34	26.99 ± 10.27	23.70 ± 7.64	25.25 ± 6.12	23.86 ± 5.15	33.78 ± 11.65	33.80 ± 14.09	36.27 ± 7.91

Table 4. The PPMCs among risk predictor variables and variables significantly affected by the main effect of height and distance.

	Subjects ($n = 6$)									
Pearson's correlation	Peak VGRF	Peak PTASF	Knee compressive force	Hip compressive force	Quadriceps muscle force	Hamstring muscle force	Gastrocnemius muscle force	Ankle flexion	Knee flexion	Hip flexion
Peak VGRF (BW)	1.00	0.33	0.95**	0.98**	0.23	−0.90*	0.15	0.50	−0.06	−0.27
Peak PTASF (BW)		1.00	0.54	0.29	0.85*	−0.20	−0.88*	−0.82*	0.19	−0.28
Knee compressive force (BW)			1.00	0.93**	0.33	0.81**	0.34	−0.60	−0.14	−0.42
Hip compressive force (BW)				1.00	0.21	0.92*	0.11	−0.48	0.10	−0.19
Quadriceps muscle force (BW)					1.00	−0.17	0.92**	0.91**	−0.31	0.25
Hamstring muscle force (BW)						1.00	−0.25	0.11	0.08	−0.40
Gastrocnemius muscle force (BW)							1.00	0.88*	−0.04	−0.08
Ankle flexion (deg)								1.00	0.32	0.13
Knee flexion (deg)									1.00	0.83*
Hip flexion (deg)										1.00

*$p < 0.05$; **$p < 0.01$.

et al. 2007) between genders during single-leg landings. Given that the single-leg landing tasks in the current study were sagittal plane dominant possibly explains the lack of kinematic differences between genders, whereby these differences may become more pronounced during out-of-plane movements. Out-of-plane motion such as side-step cutting has been shown to lead to valgus collapse and subsequently ACL injury (Boden et al. 2000; Teitz 2001; Olsen et al. 2004; Krosshaug et al. 2007); however, it is not clear whether valgus collapse causes ACL injury or occurs as a result of the ACL being torn (Olsen et al. 2004). In this study, visual inspection of the body's COM trajectory in the frontal plane for all landing configurations revealed an almost linear trend suggesting little out-of-plane motion. Nonetheless, an area of future research remains single-leg landing from increasing vertical heights and horizontal distances that involves out-of-plane motion such as landing to the medial or lateral aspect of the knee. It is also possible that given the current study was performed under controlled laboratory settings without the demands of a competitive sport situation, females landed in a way that was protective of their ACL. Previous studies (Devita and Skelly 1992; Zhang et al. 2000) have reported that internal and external forces at the lower extremity joints can be modulated by body kinematics. The body needs to be in a position that allows the muscle to absorb GRFs. The literature investigating single-leg landings reported that females experienced higher VGRF (Pappas et al. 2007; Schmitz et al. 2007) or no difference in VGRF (Lephart et al. 2002) compared to males. This study does not support these findings and revealed that females had significantly lower peak VGRF compared to males, which contradicted our hypothesis. Our findings showed that for many of the single-leg landing configurations, females had higher quadriceps to hamstring ratios compared to males, which is in general agreement with the literature (Colby et al. 2000; Malinzak et al. 2001; Decker et al. 2003). One possible explanation why females may have experienced lower peak VGRF is because of their higher quadriceps to hamstring ratios, which was shown by an earlier study (Podraza and White 2010) to result in a significant decrease in VGRF. Further rationalisation for the lower peak VGRF in females compared to males during single-leg landing has yet to be elucidated and requires additional research.

The results of our study suggest that compared to males, females did not exhibit significantly different quadriceps, hamstrings and gastrocnemius muscle force characteristics (Table 2), which is consistent with previous studies (Fagenbaum and Darling 2003; Pappas et al. 2007). This finding did not support our hypothesis that lower extremity muscle forces would be significantly different between genders. Even though we observed significant gender differences in hip axial compressive force, the significance of this finding and its implication for sex differences in ACL injury rates remain unclear.

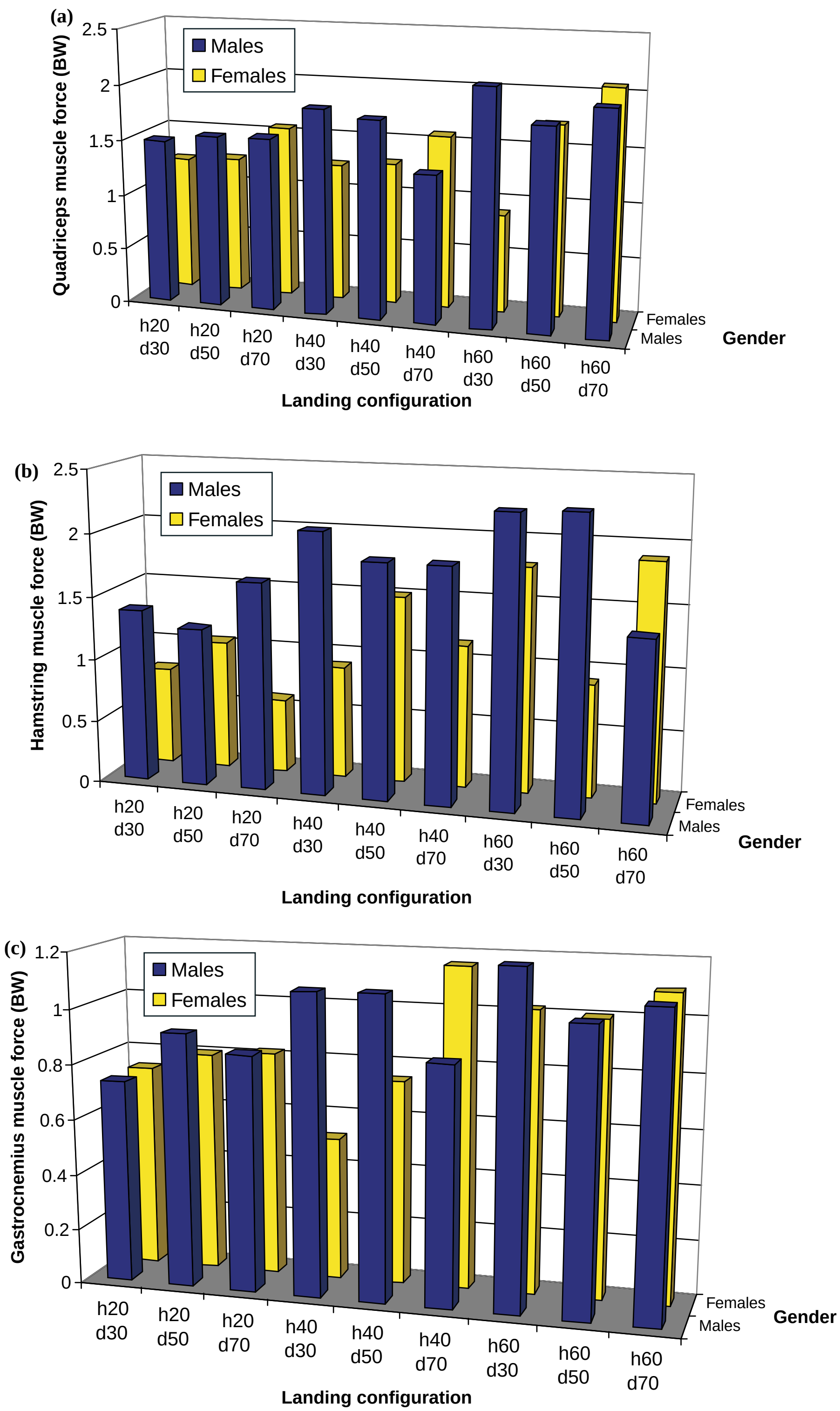

Figure 6. The average muscle force for males (blue) and females (yellow) at the nine single-leg landing configurations for (a) quadriceps, (b) hamstring and (c) gastrocnemius muscles.

Stability to the knee joint during single-leg landing is provided by the surrounding musculature. Several studies (Renstrom et al. 1986; Hewett et al. 1996; Rozzi et al. 1999; Malinzak et al. 2001) have investigated the role of ACL agonist (hamstring and gastrocnemius) and antagonist (quadriceps) on knee biomechanics. Although no statistically significant difference was observed between genders, our findings revealed that, generally speaking,

muscle forces are higher in males across all landing configurations except at landing configurations using $d = 70$ cm where males and females had approximately the same muscle force (Figure 6(a)–(c)). Furthermore, Figure 6(a)–(c) showed that muscle forces in general were the highest for the quadriceps at low vertical landing height ($h = 20$ cm), and highest for the hamstring at higher heights ($h = 40$ cm and $h = 60$ cm). Interestingly, at every height, the quadriceps muscle force increased with increasing distance for females, while this pattern did not hold for males (Figure 6(a)). This suggests perhaps that longer distance of landing is an important variable among females that should be considered in future single-leg landing studies. Finally, it was also found that the gastrocnemius showed the lowest muscle force of the three major muscles studied.

An early study (Lees 1981) demonstrated that muscular activity during jump landings can modify GRFs. Many single-leg landing studies in the literature investigating ACL injury do not include the effect of muscles, even though an *in vivo* study (Beynnon and Fleming 1998) demonstrated that an increase in quadriceps activity relative to hamstring activity can significantly increase ACL strain. Despite this, the effect of muscle forces on non-contact ACL injuries remains limited and controversial. To elucidate, an early study (McConkey 1986) is probably the first to describe eccentric quadriceps contraction as the intrinsic force responsible for ACL injury. Later studies have confirmed that increased quadriceps muscle activation may be a risk factor for non-contact ACL injuries (Draganich and Vahey 1990; Malinzak et al. 2001; Chappell et al. 2002; Markolf et al. 2004; Hewett et al. 2006; Timothy et al. 2007), and our results corroborate these findings showing that an increase in quadriceps muscle force is associated with an increase in peak PTASF. Given that PTASF was used as an indicator of risk of ACL injury, this increase in peak PTASF may result in greater loading on the primary restraint to ATT, that is, the ACL.

An early study (Solomonow et al. 1989) suggested that strength training of the hamstring muscles can help prevent injury to the ACL. A number of other studies using various methodologies support this finding showing that the hamstrings can provide a counter-balancing force to protect against quadriceps induced ATT, thereby reducing the risk of ACL injury (Draganich and Vahey 1990; O'Connor 1993; Liu and Maitland 2000). Our results showed that hamstring muscle force was significantly and negatively correlated with the peak VGRF. Other studies have shown that compressive loading at the tibial–femoral joint acts to press together the articular surfaces, thereby limiting ATT and consequently limiting strain at the ACL (Torzilli et al. 1994; Fleming et al. 1999). Our results revealed that hamstring muscle forces significantly and positively correlated to both knee and hip joint axial compressive forces. These results combined together suggest that increased hamstring muscle force has the potential to reduce the risk of non-contact ACL injury.

A cadaveric study (Durselen et al. 1995) and an analytical study (Shelburne and Pandy 1997) both demonstrated that contraction of the gastrocnemius muscles did not strain the ACL over the entire range of knee flexion. Both the results of *in vivo* studies (Fleming et al. 2001a, 2001b) and an analytical study (Pflum et al. 2004) contradicted these findings by demonstrating that the gastrocnemius muscle was an antagonist to the ACL. Our results are not consistent with these studies and support the notion that gastrocnemius muscle can protect the ACL.

Body kinematics can play a role in terms of its effect on non-contact ACL injury risk. Our results showed that ankle flexion angle was significantly correlated to peak PTASF. These results indicate the potential of increased ankle flexion angle to modulate peak PTASF and subsequent risk to non-contact ACL injury. This finding is in agreement with the literature that showed that the athletes who injured their ACL had smaller ankle flexion angles (McNitt-Gray 1993; Self and Paine 2001; Madigan and Pidcoe 2003; Boden et al. 2009; Shimokochi et al. 2009).

A limitation of this study is that the MSMs applied in this study did not include the four major knee ligaments but instead modeled the knee as a revolute joint. The literature implicates that these ligaments aid knee stability by limiting the knee's range of motion (Beynnon et al. 1997). The MSMs used in this study also did not incorporate the subject-specific tissue characteristics such as tibial plateau geometry and femoral intercondylar notch width which are implicated as risk factors to non-contact ACL injury. Given the limitations in model anatomy, it is not expected that the MSM resembles the human body of the subjects tested and therefore the MSMs in this study should be extended to include accurate anatomical geometries of each subject so as to include factors such as tibial plateau geometry and femoral intercondylar notch width, which are implicated to increase the risk of non-contact ACL injury. Another limitation is that the human body was modelled as a rigid structure without including all the detailed deformable structures of the knee joint. This implies that the results need to be interpreted with this in mind as no damping characteristics of the human body were considered. According to a study (Gruber et al. 1998), the joint moments can be very high for rigid body models compared with wobbling mass models. It can also be argued that the findings of the current study may be limited, given that there is equally a limited number of studies supporting the use of peak VGRF and peak PTASF as ACL injury risk predictor variables. A further limitation of this study is that the joint reaction and muscle forces during single-leg landings could not be directly validated. As well, a recent study investigating side-step cutting (Kristianslund et al. 2012) has shown low-pass filtering of the kinematic and force plate data at different cut-off frequencies as done in this

study can result in inaccuracies. Within these limitations, the current study assumes that the applied MSMs are a good starting point for estimating the joint reaction and muscle forces during single-leg landings.

Given the number of variables that can affect the ACL loads *in vivo* and the small sample size may have overlooked some significant differences, our results revealed little gender differences in the musculoskeletal variables investigated and suggest that other factors such as upper body kinematics, out-of-plane kinematics, anatomy, hormones, fatigue or a combination of all these may perhaps better explain the gender bias in non-contact ACL injury rate during single-leg landing. Perhaps the risk of non-contact ACL injury is likely multifactorial with no single causative factor being solely responsible for the gender bias in non-contact ACL injury rate. Perhaps a new multi-factorial study approach capable of including all risk factors implicated to increase the risk of non-contact ACL injury into a single unified study environment is needed. Or perhaps the lack of gender differences in lower extremity kinematics and muscle forces in this study may be attributed to other force-absorbing compensatory mechanisms that were not included in this study. Therefore, we conclude that factors other than those evaluated in this study need to be considered when attempting to determine the reasons underlying the gender bias in non-contact ACL injury rate. While the findings stemming from this study require further corroboration, its ramifications relative to single-leg landing remain tenable.

Available for download

The presented model can be downloaded from: http://forum.anyscript.org/.

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Conflict of interest statement

There are no conflicts of interest in this study from any of the authors.

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